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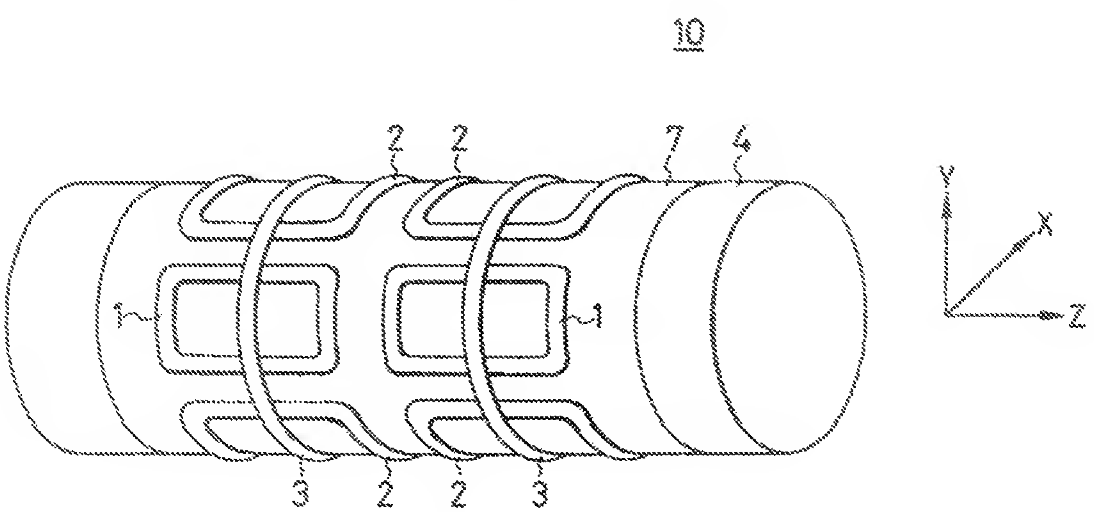
(54) **Magnetic field coil for NMR-CT**

(57) A magnetic field generating coil structure for use in a nuclear magnetic resonance computer tomograph in which the boundaries of a slice are precisely determined due to elimination of magnetic coupling between an RF coil and a gradient field coil. The gradient field coil includes coils 1, 2 and 3 for generating magnetic fields with gradients in X, Y and Z orthogonal directions while a uniform field generating structure produces a uniform field in the Z direction. An RF coil structure is disposed within the uniform field generating coil structure and the gradient field generating coil structure. In accordance with the invention, a cylindrical shielding layer (7) made of an electrically conductive material is disposed on at least one of the inside and outside of the gradient field generating coil structure. The thickness of the shielding layer is made equal to or larger than a skin depth determined by the fequency of the RF field and the resistivity of the shielding layer and made smaller than a skin depth determined by the gradient field control frequency and the resistivity of the shielding layer.

FIG. 7



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The drawing(s) originally filed was/were informal and the print here reproduced is taken from a later filed formal copy.

FIG. 1

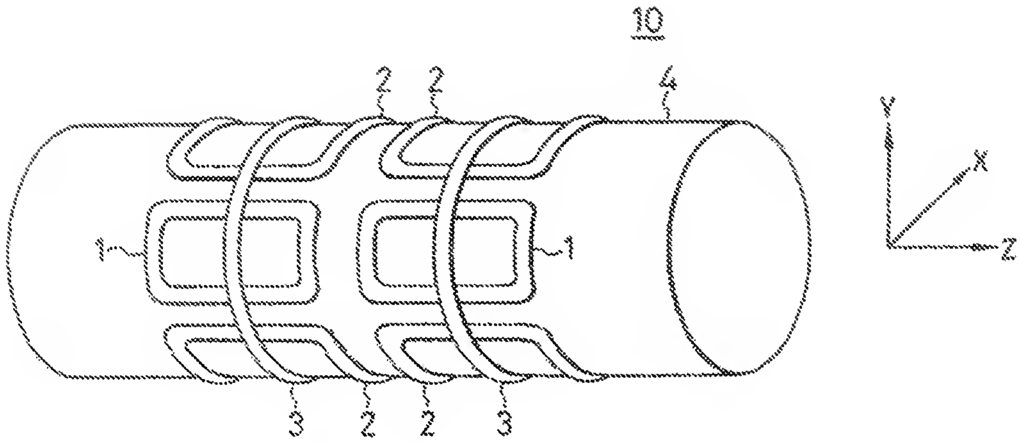


FIG. 2

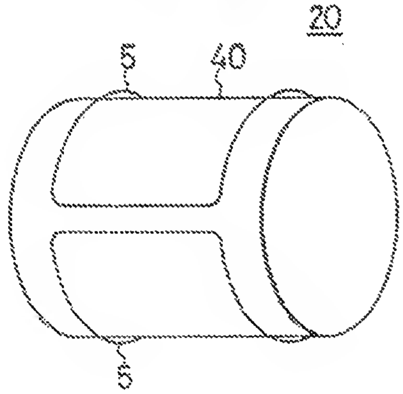


FIG. 3

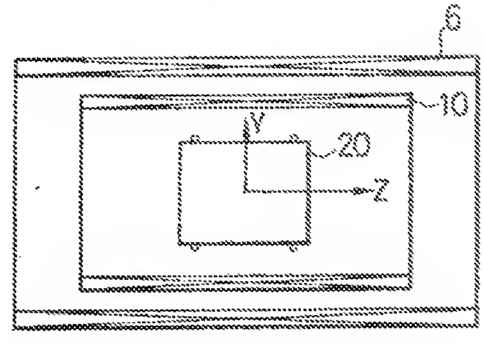


FIG. 4

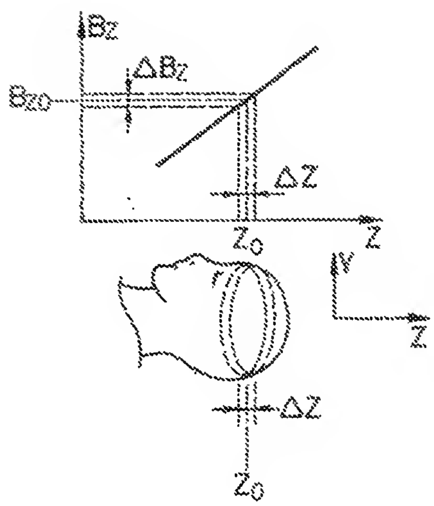


FIG. 5

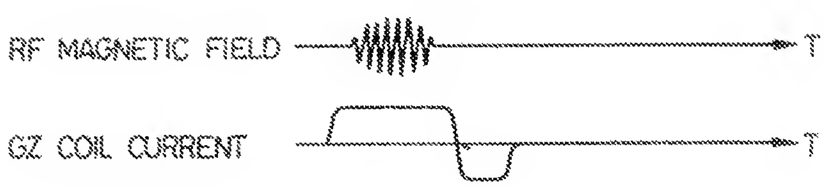


FIG. 6

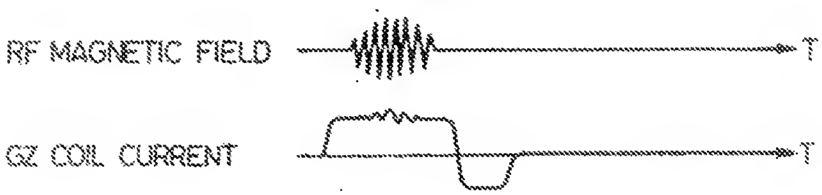


FIG. 7

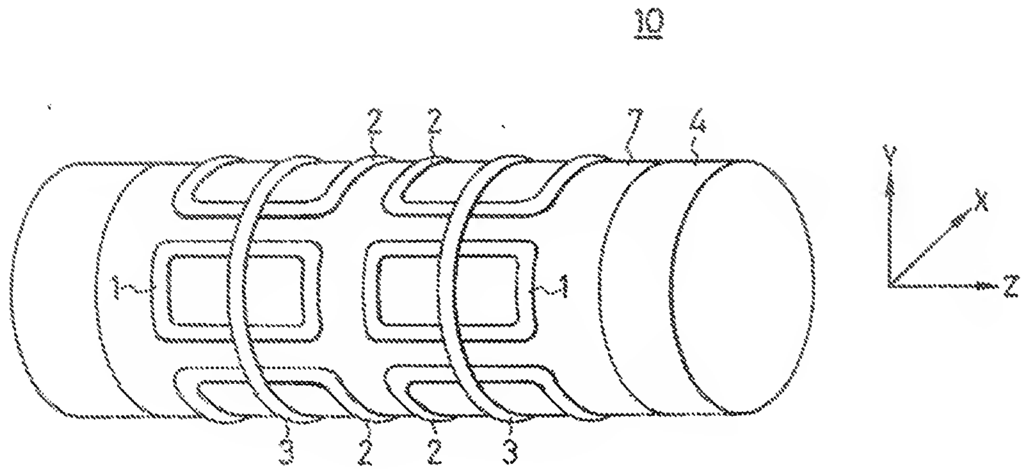


FIG. 8

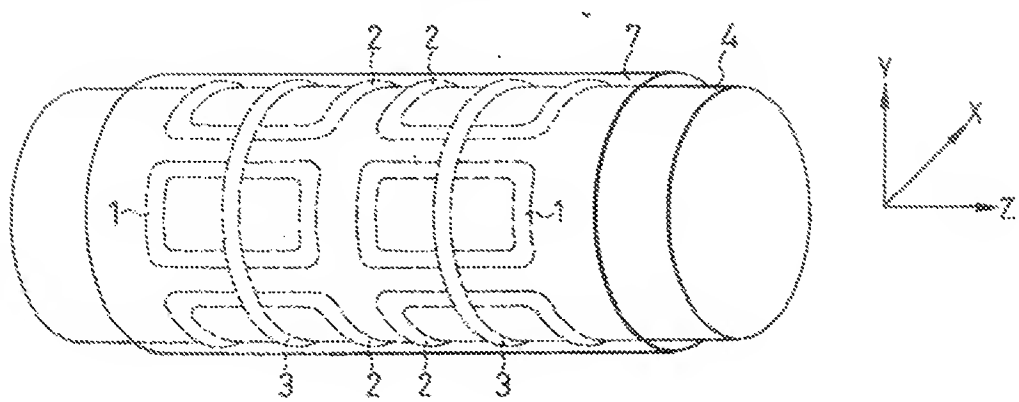


FIG. 9

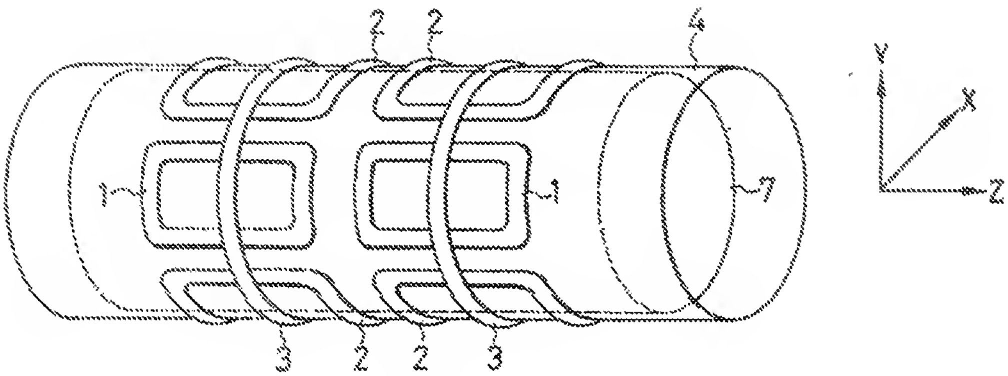
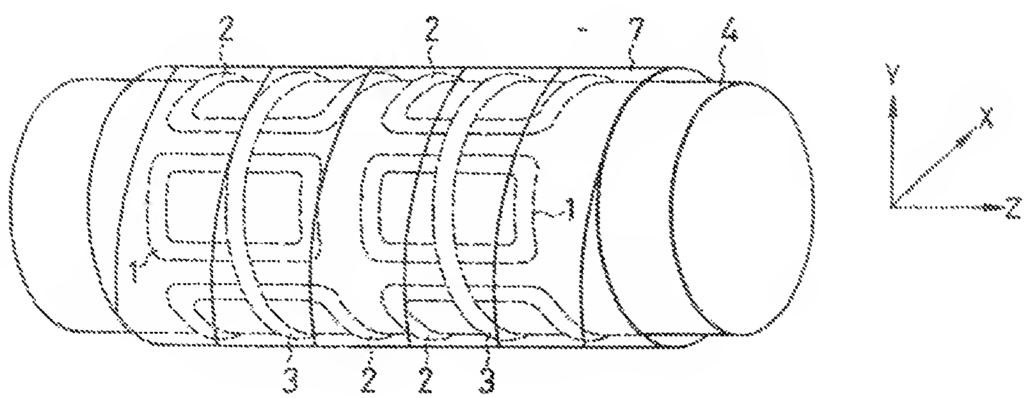


FIG. 10



SPECIFICATION

Magnetic field coil for NMR–CT

The present invention relates to a magnetic field coil structure of a nuclear magnetic resonance apparatus for a computer tomograph (referred to hereinafter as an NMR–CT), and particularly to an NMR–CT having improved reliability.

Figs. 1 to 3 show an example of a conventional apparatus of this type, of which Fig. 1 is a perspective view of a conventional gradient field coil structure, Fig. 2 is a perspective view of an RF coil structure, and Fig. 3 shows the construction of a magnetic coil for the NMR–CT, the latter being composed of the gradient field coil structure and the RF coil structure. In these figures, the gradient field coil 10 is constructed with a gradient field G_x coil composed of four coils, supported by a coil support frame 4, which generate the magnetic field with the gradient in the X direction, a gradient field G_y coil 2 composed of four coils which generate the magnetic field with the gradient in the Y direction (orthogonal to the X direction), and a gradient field G_z coil 3 composed of two coils for generating the magnetic field with the gradient in the Z direction (orthogonal to the X and Y directions). A magnetic field with the gradient in an arbitrary direction can be produced by suitably controlling the currents flowing through the G_x coil 1, G_y coil 2 and G_z coil 3.

As shown in Fig. 2, inside the coil structure 10 there is disposed an RF coil 20. With reference to Fig. 3, the RF coil 20 is composed of a cylindrical support frame 40 with a saddle coil 5 disposed thereon. A uniform field coil structure 6 is further provided for producing a highly uniform magnetic field in the Z direction. Although the uniform field coil structure 6 is shown in Fig. 3 as being disposed outside the gradient field coil structure 10, the coil structures 6 and 10 may be interchanged in position if desired.

In operation, in an NMR imaging system, the body portion to be examined is disposed inside the RF coil structure 20 to obtain a NMR signal therefrom. Fig. 4 illustrates the determination of a tomographic plane through a human head. In Fig. 4, a solid line shows the relation of the position of the head to a magnetic field of intensity B_z in the Z direction, which is a sum of a gradient field and a highly uniform field.

Assuming a slice of the head having a center line at Z_0 and a thickness of ΔZ, the intensity of the magnetic field in which the slice is disposed is between $(B_{z0} - \frac{1}{2}\Delta B_z)$ and $(B_{z0} + \frac{1}{2}\Delta B_z)$. Therefore, the resonance frequency of a substance (protons) in the slice from which the NMR signal is obtained falls within a range from $\gamma(B_{z0} - \frac{1}{2}\Delta B_z)$ to $\gamma(B_{z0} + \frac{1}{2}\Delta B_z)$, where γ is the gyromagnetic ratio. When a high frequency (RF) magnetic field generated by the RF coil structure 20 and whose frequency components are within the above range is applied to the slice additionally, only protons in the slice are excited and absorb energy. An NMR detection signal is accordingly produced from which an NMR tomogram can be reconstructed by processing with a computer. As mentioned above, in order to determine a certain slice plane of the head, it is necessary to apply both an RF field of a frequency in the constant range and a gradient field to the head or other body part being examined.

In order to generate an RF field whose frequency range is from $\gamma(B_{z0} - \frac{1}{2}\Delta B_z)$ to $\gamma(B_{z0} + \frac{1}{2}\Delta B_z)$, a current is supplied to the RF coil which is obtained by modulating a sinusoidal current having a frequency $\gamma(B_{z0})$. The modulation to be used is selected such that a frequency spectrum obtained by Fourier transformation of a current waveform modulated thereby falls within the frequency range of $\gamma(B_{z0} - \frac{1}{2}\Delta B_z)$ to $\gamma(B_{z0} + \frac{1}{2}\Delta B_z)$.

Fig. 5 shows the RF field and an example of the waveform of a current flowing through the G_z coil for producing the gradient firld in the Z direction. The abscissa indicates time T. In Fig. 5, the G_z coil current is inverted after application of the RF field so that the phase of the excited protons is made uniform.

The gradient field coil structure 10 shown in Fig. 1 and the RF coil structure 20 shown in Fig. are coaxially arranged as shown in Fig. 3. If the coil structures 10 and 20 were manufactured and arranged ideally, there would be no electromagnetic coupling therebetween, which is the case for the waveforms shown in Fig. 5. However, it is impossible as a practical matter to manufacture and arrange them ideally, resulting in unavoidable electromagnetic coupling therebetween.

An effect of such electromagnetic coupling is shown in Fig. 6 which shows actual waveforms obtained by a conventional magnetic field coil structure for an NMR–CT. This is, when the RF coil is energized in a pulsed manner during the energization of the G_z coil, an RF pulse voltage is induced in the G_z coil, due tio which the G_z coil current varies. Due to the variation of the G_z coil current, an gradient field which is in a range of $(B_{z0} - \frac{1}{2}\Delta B_z)$ to $(B_{z0} + \frac{1}{2}\Delta B_z)$ is applied also to body portions other than the slice of thickness ΔZ centered at Z_0. That is, protons outside the slice are also excited and the obtained NMR signal contains information from not only the desired slice but also from portions around the slice. This results in blurring of the tomogram.

An object of the present invention is thus to provide a magnetic field coil structure for an NMR–CT which has an RF coil and a gradient field coil having no high frequency electromagnetic coupling therebetween, and which consequently provides exact slice plane deter-

mination.

According to the present invention, the above object is achieved by the provision of a magnetic field coil structure for an NMR–CT having a shielding layer made of an electrically conductive material in the form of a cylinder disposed at least inside or outside the gradient field coil.

In the accompanying drawings:

Figure 1 is a perspective view of a conventional gradient field generating coil;

Figure 2 is a perspective view of an RF field generating coil;

Figure 3 is a cross-sectional view of a magnetic field coil structure for an NMR–CT;

Figure 4 illustrates the principles of determining a slice plane for NMR tomography;

Figure 5 shows ideal waveforms of the RF field and current flowing through the gradient coil;

Figure 6 shows similar waveforms to those of Fig. 5 for an actual case;

Figure 7 is a perspective view of a gradient coil constructed according to a preferred embodiment of the present invention; and

Figure 8, 9 and 10 perspective views of gradient coils according to other embodiments of the present invention.

In Fig. 7, which is a perspective view of a gradient field generating coil structure 10 constructed according to a preferred embodiment of the present invention, a cylindrical shielding layer 7 made of electrically conductive material is disposed around a cylindrical coil support frame 4, and gradient field generating coils 1, 2 and 3 are provided on the outer surface of the shielding layer 7.

The thickness t of the shielding layer 7 is selected so as to satisfy the following:

$\delta(f_{RF}) \leqq t < < \delta(f_G)$,

where f_{RF} is the frequency of the RF magnetic field, f_G is a control frequency for the gradient magnetic field generating coil structure 10, and $\delta(f)$ is a skin depth which is represented by

$$\frac{2\rho}{2\pi f\mu},$$

where ρ is the resistivity of the shielding layer, f is frequency, and μ is the permeability of the shielding layer.

That is, when the thickness of the shielding layer is made equal to or greater than the skin depth determined by the RF field frequency and the resistivity of the shielding layer but smaller than the skin depth determined by the gradient field control frequency and the resistivity of the shielding layers, the RF field generated by an RF coil 20 and penetrating the ramp field coil 10 is shielded by eddy currents produced in the shielding layer 7, and thus the variation of the gradient field coil current due to the RF field is eliminated. Therefore, it becomes possible to precisely determine the slice bounderies. On the other hand, since the thickness t of the shielding layer 7 is sufficiently smaller than the skin depth $\delta(f_G)$ associated with the gradient field coil control frequency, the gradient field is not shielded electromagnetically by the layer 7.

For values of:

$f_{RF} = 10$ MHz,
$f_G = 1$ KHz
$\rho = 2.7 \times 10^{-8}\ \Omega m$
$\mu = 4\pi \times 10^{-7}$ H/m
$\delta(_{RF}) = 27\ \mu m$ and
$\delta(f_G) = 2.6$ mm,

$\delta(f_{RF})$ is determined as:

$$\delta(f_{RF}) = 27\ \mu m \leqq t < < \delta(f_G) = 2.6 \text{ mm}.$$

Thus, a suitable value for $\delta(f_{RF})$ is, for example, 30 μm.

Although, in the above embodiment, the shielding layer 7 is disposed inside of the gradient field coil structure 10, it is possible to dispose it outside of the gradient field coil structure 10, as shown in Fig. 8, or inside of the coil support frame 4, as shown in Fig. 9. Moreover, if shielding layers are disposed in both sides of the gradient field coil structure, the shielding effect may be further improved.

Fig. 10 is a perspective view of a ramp field coil structure according to another embodiment of the present invention. In Fig. 10, a shielding layer 7 is formed on the gradient field coil structure by winding thereon a thin conductive tape having desired thickness. Adjacent edges of turns of the tape are overlapped. It is possible to use a conductive tape having a thickness smaller than the skin depth $\delta(f_{RF})$ and to overlappingly wind it on the gradient field coil structure to form a multilayered shielding layer having a thickness equal to or greater than the skin depth $\delta(f_{RF})$. A metal mesh may be used to form the shield layer 7.

As mentioned hereinbefore the cylindrical shielding layer provided on at least one side of the gradient field coil structure ensures that high frequency electromagnetic coupling between the RF coil and the gradient coil is eliminated, resulting in precise determination of the slice plane.

CLAIMS

1. A magnetic field generating coil structure for use in an NMR–CT, comprising: a gradient field generating coil structure composed of field generating coil means for generating the magnetic field with the gradient in an X direction, gradient field generating coil means for generating the magnetic field with the gradient in a Y direction orthogonal to the X direction, and gradient field generating coil

means for generating the magnetic field with the gradient in a Z direction orthogonal to the X and Y directions, a uniform field generating structure for generating a uniform field in the Z direction; an RF coil structure disposed within said uniform field generating coil structure and said gradient field generating coil structure; and a cylindrical shielding layer made of an electrically conductive material disposed on at least one of an inside and outside of said gradient field generating coil structure.

2. The magnetic coil structure as claimed in claim 1, wherein said shielding layer is circular in cross section.

3. The magnetic coil structure as claimed in claim 1, wherein said shielding layer is formed by winding a conductive tape with edge portions of adjacent turns overlapping.

4. The magnetic coil structure as claimed in claim 1, wherein said shielding layer comprises a metal mesh.

5. The magnetic coil structure as claimed in claim 1, wherein a thickness of said shielding layer is equal to or larger than a skin depth determined by an RF field frequency and resistivity of said shielding layer and smaller than a skin depth determined by a gradient field control frequency and the resistivity of said shielding layer.

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